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## Finite Element Analysis of Mechanical Behaviors of Coronary Stent

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### Abstract

One of the most prevalent health problems is coronary artery disease which may cause pain and heart attack. Stent implantation is a non-surgical method to treat the coronary artery disease that can support arterial walls and reduce the risk of heart attack. Utilizing finite element analysis to study the mechanical characteristics of this device is an efficient way to modify the design of stent and its performance. The main aim of the present work is to investigate the expansion characteristics of a certain stent as it is deployed and implanted in an artery containing a plaque, and try to reach to a model close to a real condition of stent implantation. A bi-linear elasto-plastic material model for stent and hyperelastic material models for balloon, artery, and plaque have been assumed. This model includes the internal pressure of blood. Stress distribution, outer diameter changes, and bending are investigated. A commercially available software has been used.

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**Keywords:** Coronary Stent; Balloon; Plaque and vessel; Finite element method; Nonlinearity

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### 1. Introduction

In western countries, coronary heart disease is the most common reason for death. Coronary artery disease is specific to the arteries of the heart. Coronary artery disease, also known as atherosclerosis, occurs when excess cholesterol attaches itself to the walls of the blood vessels. Embedded cholesterol also attracts cellular waste products, calcium and fibrin. This leads to a thickening of the vessel wall by complex interaction with constituents of the artery. The resulting pasty build-up known as plaque can

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narrow or even block an artery [1]. Several procedures are available to revascularise an occluded artery, including balloon angioplasty and stenting, bypass surgery and atherectomy [2].

Stenting shows some advantages compared to other possible treatments, as it does not require any surgical operation and has less complication, pain and a more rapid recovery [3]. So the use of coronary stents in interventional procedures has rapidly increased from 10% in 1994 to over 80% in current practice [4]. An estimated 652,000 percutaneous coronary interventions involving stents deployments were performed in the United States in 2006 [5].

Over 100 different stent designs are currently being marketed or are in evaluation for vascular and non-vascular indications [6]. In order to have the better output of stent implantation, it is needed to analyze the mechanical behavior of the stent before manufacturing and utilizing. One of the most effective methods to investigate the mechanical behavior of the stent is finite element method. In comparison with expensive experiments carried out in hospitals and laboratories, numerical simulations accomplished by computers have advantages in both flexibility and cost [7]. In the early works of simulation, researches used single stent models [8]. Later on, researchers tried to have a better and even more precise analyses, therefore, they included balloon in their models [1, 9-11]. Recently, in order to reach to a more real analysis of stent implantation, researchers try to model vessel and plaque as well [12-14]. The main objective of this paper is to present a model that is more accurate and closer to the real condition of stent implantation by such a modeling that contains internal blood pressure, balloon, stent, plaque and vessel.

## 2. Finite Element Model

In this section, modeling of various parts used in analysis of mechanical behavior of Coronary Stent is presented. This part includes the modeling of stent, balloon, and vessel with plaque.

### 2.1. Stent

A balloon-expandable Palmaz-Schatz stent was modeled in this study. Primary model of stent was produced using commercially available software. The simulation was carried out to expand a stainless steel 304 stent of 3mm outer diameter and 2.9mm inner diameter with 10mm in length.

A bi-linear elasto-plastic material model was assumed for the stent material. The material properties were chosen to approximately represent stainless steel 304 and are [1]: Young's modulus=193GPa; shear modulus= $75 \times 10^6$ MPa; tangent modulus=692MPa; density= $7.86 \times 10^{-6}$ kg/mm<sup>3</sup>; yield strength=207MPa; Poisson's ratio=0.27.

The stent was modeled by linear 8-nodes three dimensional block elements with 6 degrees of freedoms at each node. The finite element model of the stent consisted of a total of 2700 elements. Fig 1a shows the finite element model of the stent.

### 2.2. Balloon

The balloon as a medium to expand the stent was modeled to be 12mm in length. The outer diameter and the thickness of the balloon were 2.9 mm and 0.1 mm, respectively. A polyurethane rubber type material was used to represent the balloon. Polyurethane is incompressible and was modeled by using a nonlinear first-order Mooney–Rivlin hyperelastic constitutive equation.

The energy function's constants used are:  $C(10)=1.06881$ MPa,  $C(01)=0.710918$ MPa and  $d=1070$ kg/m<sup>3</sup>. Where  $C(10)$  and  $C(01)$  are the energy function invariants, and  $d$  is the density of the polyurethane [1, 10].

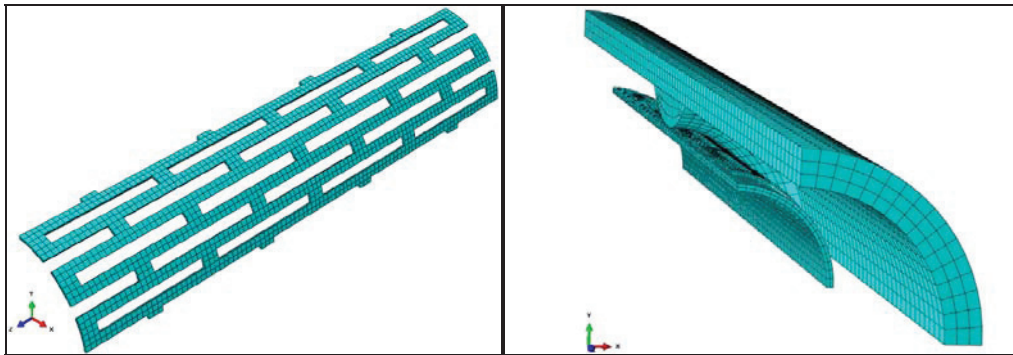


Fig. 1. (a) Finite element model of one quarter of the stent; (b) Balloon-stent-vessel model.

Similar to the stent, the balloon was modeled by linear 8-nodes three dimensional block elements with 6 degrees of freedoms at each node. The finite element model of the balloon consisted of a total of 3600 elements.

### 2.3. Vessel with plaque

With the assumption of homogenous and isotropic material, the coronary artery was modeled as an idealized vessel. The geometrical properties of the vessel and plaque are as follows:

Vessel's length=20mm; vessel's inner diameter=4mm; vessel's outer diameter=5mm; plaque's length=3mm; plaque's inner diameter=3mm.

The two materials of the vessel and plaque were modeled using a third-order Mooney–Rivlin hyperelastic constitutive equation (Eq. 1):

$$W = a_{10}(I_1 - 3) + a_{01}(I_2 - 3) + a_{20}(I_1 - 3)^2 + a_{11}(I_1 - 3)(I_2 - 3) + a_{30}(I_1 - 3)^3 \quad (1)$$

where  $W$  is the strain-energy density function of the hyperelastic material,  $I_1$ ,  $I_2$  and  $I_3$  are the strain invariants and  $a_{ij}$  are hyperelastic constants. Table 1 summarizes the constants used for the hyperelastic constitutive equations of the two material models [15].

Table 1. Hyperelastic constants to describe the arterial tissue and stenotic plaque non-linear elastic behavior.

	Arterial wall tissue (kPa)	Stenotic plaque tissue (kPa)
$a_{10}$	18.90	-495.96
$a_{01}$	2.75	506.61
$a_{20}$	85.72	1193.53
$a_{11}$	590.43	3637.80
$a_{30}$	0	4737.25

The vessel and plaque were modeled by using of 8-nodes three dimensional block elements with 6 degrees of freedoms at each node. The vessel and the plaque consisted of 3840 and 560 elements, respectively. Fig 1b shows the balloon-stent-vessel model.

### 3. Loading and Solution

Due to the symmetrical conditions, only a quarter of the model was used to simulate the expansion process, as shown in Fig 1b. Symmetric boundary conditions were imposed on the nodes of the balloon, stent and vessel in the planes of symmetry where all the nodes perpendicular to axis-y were not allowed to move in direction-y and all the nodes perpendicular to axis-x were not allowed to move in direction-x. In addition, for balloon, the nodes located at the two ends of the balloon are fully fixed, and for vessel, only the movement in radial direction was permitted for the nodes located at the two ends of the vessel, and the plaque was attached to the balloon.

An automatic surface to surface algorithm approach available in software was selected in order to cope with the nonlinear contact problem among the surfaces.

The loading process consisted of two steps. In the first step, without considering the existence of the balloon and the stent, a constant internal pressure equal to 13.3kPa was applied to the vessel and the plaque. This pressure is equal to the blood pressure of 100mmHg [15]. The pressure simulates the internal pressure of the blood and causes the vessel to expand, and also, induces an initial stress. This step causes the modeled vessel and the plaque to be as close to the reality as possible. In the second step, by keeping the initial pressure applied to the vessel and plaque, a constant pressure was applied to the internal surface of the balloon. This pressure was applied with a constant rate in 1.635 seconds and its value was varied from 0 to 0.4MPa.

### 4. Results and Discussion

In this section, the results of the finite element analysis of the coronary stent are presented. The results include stress distribution, outer diameter changes and bending.

#### 4.1. Stress distribution

The distribution of von Mises stress in the expanded stent is shown in Fig 2a. As can be seen in this figure, the regions of high stress are located at the four corners of the cells. This is because of the struts being pulled apart from each other to form a rhomboid shape of cells during the expansion. The value of maximum von Mises stress in the stent is 257.5MPa.

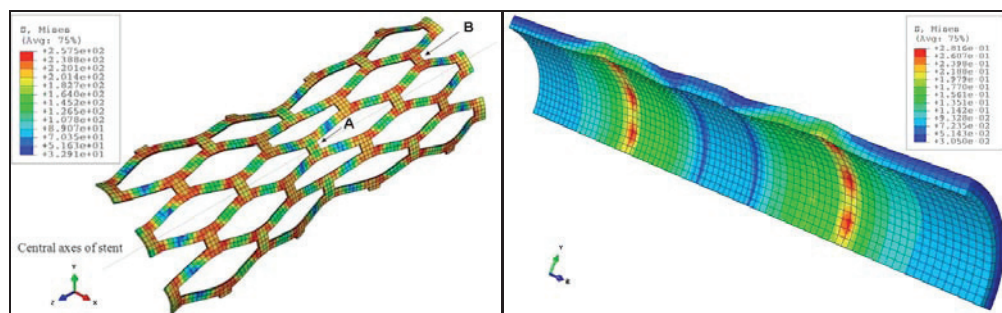


Fig. 2. (a) Distribution of von Mises in the stent; (b) Distribution of von Mises in the vessel at maximum expansion.

As expected, symmetrical stress distributions around the central axis of stent can be seen in Fig 2a that verifies the validation of modeling and imposed boundary conditions. Moreover, in comparison with

analysis in [1] which shows the maximum stress of 249MPa, it has been observed that the obtained maximum stress in present work (i.e. 257.5MPa) is acceptable. The distribution of von Mises stress in the expanded vessel is shown in Fig 2b. As seen in this figure, the regions of high stress are located where the stent bending occurred. The value of maximum von Mises stress in the vessel is 0.282MPa. This might lead to a possible damage to the vessel at these critical points shown in Fig 2b.

#### 4.2. Outer diameter changes

Fig 3a shows the outer diameter of the stent against the expanding pressure. As can be seen in Fig 2a, because of the existence of the plaque and the pressure applied by it, different positions of stent have different diameters. Here, only the outer diameter changes of points A and B (as shown in Fig 2a) were derived and shown.

Figure 3a shows that the rate of increment of stent diameter at points A and B is almost identical as pressure changes from 0MPa to 0.25MPa. From the pressure 0.25MPa, because of the contact between point A and the plaque, stent diameter increases with a low rate at point A, while at point B, the rate of increment of the diameter increases significantly until the pressure reaches to 0.37MPa. Finally, for pressure larger than 0.37MPa, because of the contact between the stent and the vessel, the variation of diameter at point B becomes small.

#### 4.3. Bending

When the stent expands, it bends on edges in a way that the diameter at the end sides becomes larger than that of the middle of the stent. This phenomenon is called “bending” which its magnitude is equal to the difference between the diameters at the end side and the point B on stent shown in Fig 2a.

Fig 3b shows the relation between the bending and the expanding pressure. This figure shows that in pressures of 0.25MPa and 0.37MPa, the rate of bending is changed because of the contact of stent to the plaque and vessel.

The bending in stent implantation has both advantages and disadvantages. On one hand, the bending of stent causes high stress on vessel as shown as Fig 2b. These critical points might damage vessel. On the other hand, the bending can help to keep stent stable against the impact of pulsed motion of blood. Thus, study of bending duration stent implantation can be very useful to modify design of stents.

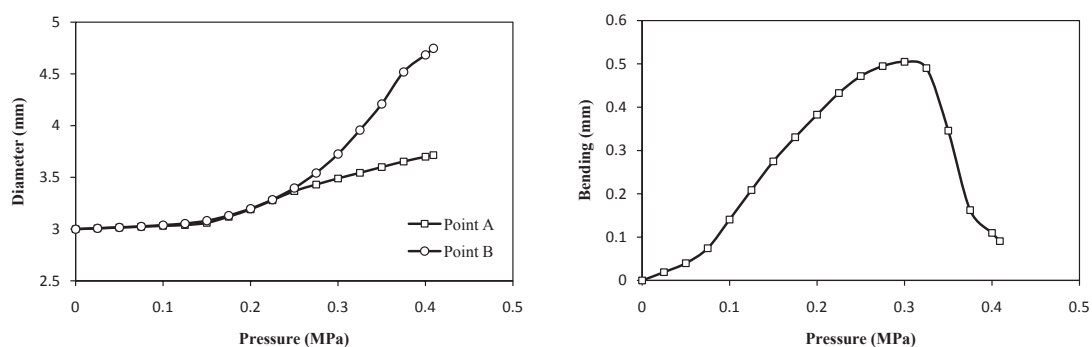


Fig. 3. The relation between (a) the outer diameter of the stent and the expanding pressure; (b) the bending and the expanding pressure.

## 5. Conclusions

The paper presents a methodology for modeling the expansion of coronary stents used in the treatment of blood vessel stenosis. In order to model much closer to the real condition, this model includes internal pressure of blood, balloon, stent, plaque and vessel. According to the analysis, the possibility of failure at the four corners of the cells is more than the other places, as these are the regions with maximum stresses. Furthermore, due to the bending at the end sides of the stent, the maximum stress on vessel occurred at the two ends. Thus, using this kind of stent would tend to damage vessel at these points. On the other hand, this bending might help stent resist against moving after implantation, therefore, more investigations on effects of bending are recommended. The analysis performed and the results obtained could be used in design and optimization of geometrical and material properties of the stents as well as in evaluation of their performance at real conditions.

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